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Patents Act 1990

ORIGINAL

PROVISIONAL SPECIFICATION

MULTIRATE COCHLEAR STIMULATION STRATEGY AND APPARATUS

The invention is described in the following statement:

MULTIRATE COCHLEAR STIMULATION STRATEGY AND APPARATUS

Field Of The Invention

The present invention relates to cochlear implant prostheses and in particular to an apparatus and method for applying stimulation to the neural structures of a cochlea in order to improve a subject's pitch perception.

Background And Summary Of Prior Art

Cochlear implant systems are used to aid patients having a hearing deficiency. More particularly, these systems include a microphone receiving ambient sounds and converting the sounds into corresponding electrical signals, signal processing means for processing the electrical signals and generating cochlear stimulating signals and an electrode assembly for applying the cochlea stimulating signals to the cochlea of an implantee. In response to the stimulations a perception of corresponding ambient sound is elicited in the implantee.

The inner ear of a normally hearing person includes hair cells which convert the displacement of the ear's basilar membrane in response to sound into nervous impulses. Different parts of the basilar membrane of the normal cochlea are displaced maximally by different frequencies of sound so that low frequency sounds maximally displace apical portions whereas higher frequency sounds cause displacement of more basal portions of the membrane. The nervous system is arranged so that a nervous impulse originating from a hair cell located adjacent an apical area of the membrane is perceived as a low frequency sound whereas a nervous impulse originating from a hair cell located adjacent a more basal position of the membrane is perceived as a higher frequency sound. This mapping of position to pitch is well known in the art as the tonotopic arrangement of the cochlea.

In a dysfunctional ear the hair cells may be damaged or absent so that no nervous impulses are generated. In such cases electrical stimulation impulses must be provided artificially to simulate the nervous activity of the hair cells in order to create a perception of sound. With reference to Figure 1 there is shown, somewhat schematically, a totally implanted cochlear implant positioned for use

by a patient. Ambient sounds are transduced by implanted microphone 103 and the transduced signal passed to electronic circuitry housed within casing 105. The transduced signal is processed according to one of many strategies, a few of which will shortly be discussed, and on the basis of the processing stimulation currents are driven between electrodes of a coupled array . For example, in "monopolar" mode stimulation currents may be forced to flow between an electrode of the electrode array 109 and extracochlear electrode 115. Nervous discharges elicited in the basilar membrane in response to applied stimulations are conveyed to the central nervous system of the wearer by auditory nerve 113. In the event that the stimulation current flows between an apical electrode such as electrode 111 and extracochlear electrode 115 then a lower pitched hearing sensation will be perceived by a wearer of the prosthesis than will be the case if stimulation current flows between basal electrode 107 and extracochlear electrode 115 because of the previously mentioned tonotopic arrangement of the cochlea. Further pitch information may be transmitted to the wearer corresponding to the rate at which stimulations are delivered.

Very many possibilities exist as to the manner in which the signal processing means operates upon the electrical signals in order to produce stimulation signals. However it has been noted in the past that simultaneous stimulation of electrodes is not generally conducive to eliciting a perception of sound that is faithful to the actual acoustic signals being processed. Consequently most cochlear prosthesis stimulation strategies stimulate by means of only one electrode at a time.

In the past designers of cochlear implant stimulation strategies have striven to optimise the intelligibility of spoken words as perceived by the wearer of a cochlear implant.

One of the earliest sound processing and cochlear stimulation strategies is described in US patent 4,532,930 to the present applicant. In that patent there is taught the use of a filter (F0) dedicated to extracting the voice pitch of a speech signal. The periodicity of the voice pitch is used to set the stimulation periodicity for two or three electrodes. A second, and possibly third, channel is analysed to determine periodicity and amplitude in a selected frequency band.

The periodicity extracted from the second filter, and possibly third filter, is used to select which electrode is stimulated for the second and third channels while the periodicity of stimulation of the channel is determined by the periodicity of the output signal from the F0 filter.

Another stimulation arrangement is described in US 4,207,441. In that system there are n electrodes each coupled to one of n filters. Each electrode is stimulated once per analysis period, with an intensity corresponding to the amplitude of the corresponding filter channel. The analysis period of this system is predetermined and hence is not related to the signal on the filter outputs.

More recently in EP 0 745 363 there is described a stimulation system which takes into account the temporal behaviour of the cochlea. In an embodiment of the invention therein described a wavelet transformation is used to extract the temporal information with a view to using this information to determine the sequence of stimulation of the electrodes. The purpose of the invention is to improve the temporal resolution in response to a rapidly changing audio spectrum.

A problem that has been faced by users of cochlear implants featuring prior art stimulation schemes is that while intelligibility of spoken words is often good the user's pitch perception, and in particular perception of music, is poor. Accordingly, it is an object of the present invention to provide an apparatus and method for use in a multi-channel cochlear implant which will improve a user's perception of pitch.

Summary Of The Invention

According to the present invention there is provided a cochlear implant prosthesis of the type having a transducer for converting an acoustic signal to an electrical signal, a plurality of bandpass filtering means responsive to said electrical signal and operatively producing a plurality of bandpass filtered signals, signal processing means responsive to said plurality of bandpass filtered signals operatively generating stimulation commands, electrode driving means responsive to said stimulation commands and an electrode array coupled to said electrode driving means for operatively delivering to a user of said cochlear implant prosthesis stimulations in accordance with said stimulation commands, said signal processing means including:

a) period estimation means, responsive to said bandpass filters operatively generating periodicity signals indicative of the periodicity of a number of said plurality of filtered signals;

b) amplitude estimation means responsive to said bandpass filters operatively generating magnitude signals indicative of the magnitude of each of said plurality of filtered signals;

c) selection means responsive to said magnitude signals arranged to determine a filtered signal of said plurality of filtered signals having a greatest magnitude, said selection means generating said stimulation commands including a command to stimulate by means of an electrode operatively tonotopically best corresponding to said filtered signal, said command to stimulate further specifying a time for stimulation to occur dependent on a corresponding one of said periodicity signals.

According to a further aspect of the present invention there is provided a method of operating a cochlear implant prosthesis of the type including a plurality of bandpass filters each having a characteristic centre frequency, said filters generating a corresponding plurality of filtered signals, said prosthesis further including stimulation delivery means coupled to an electrode array, said method including the steps of :

a) in each of a number of time intervals determining the amplitude and a periodicity value for each of said plurality of filtered signals;

b) determining which of said plurality of signals has the greatest magnitude;

c) applying a stimulation current by means of an electrode of said electrode array tonotopically closest to the centre frequency of the bandpass filter producing the signal determined in step b), said stimulation current being applied during a time interval determined from the periodicity value of the signal determined in step b).

Brief Description Of The Figures

Figure 1 depicts an implanted cochlear implant device.

Figure 1A depicts a block diagram of the functional elements of an cochlear implant according to the present invention.

Figure 2 depicts a dedicated hardware version of a cochlear implant prosthesis according to the present invention.

Figure 3A is a graph of a possible output of an amplitude estimator module of Figure 2.

Figure 3B is a graph of a possible output of a period estimator module of Figure 2 of the same channel as the amplitude estimator of Figure 3A.

Figure 3C is a graph of stimulation currents generated via an electrode in accordance with the amplitude and period estimates of Figures 3A and 3B.

Figure 3D is a a graph of stimulation currents generated via a further electrode having a stimulation current occurring simultaneously with a stimulation current in the graph of Figure 3C.

Figure 3E is a graph of a single stimulation current.

Figure 4 is a flowchart of procedural steps used in implementing the present invention by means of software on an apparatus of the type depicted in Figure 1A.

Figure 5 is a flowchart of procedural sub-steps involved in one of the boxes appearing in Figure 3.

Detailed Description

With reference to Figure 1A, there is depicted a digital hardware implementation of a cochlear prosthesis according to the present invention. Sound waves are transduced by microphone 11 and the electrical signal so produced, processed by analog conditioning module 13. Conditioning module 13 includes standard circuits for pre-amplifying and low pass filtering the signal prior to its processing by analog to digital converter 15. Analog to digital converter 15 produces a 16 bit digital signal which is conveyed to microprocessor 17. Microprocessor 17 operates according to a program stored in EPROM 19. Microprocessor 17, in accordance with its program operates upon the digital signal in order to generate a sequence of stimulation commands which are delivered to switchable current source module 23. The commands delivered to the current source module specify the amplitude of the current that is to flow from one or more electrode to one or more other, the timing of the stimulation current and the mode of the stimulation.

The term "mode" here refers to the selection of electrodes between which a stimulation current is to flow. Well known stimulation modes include bipolar, monopolar and common ground.

Upon receiving commands specifying the parameters of the stimulation to be applied, switchable current source module 23 connects various electrodes of electrode array 21 to an internal controllable current source in order to generate the appropriate stimulation. The construction of a switchable current source is well known in the art and may be found in US Pat No. 4,532,930 to the present applicant.

Figure 2 depicts a dedicated hardware implementation of the invention for purposes of explanation. While Figure 2 illustrates the invention as if individual tasks performed by microprocessor 17 were embodied in dedicated hardware, it remains the case that the invention is most readily practised according to the arrangement of Figure 1A. The invention will however be explained with reference to Figure 2 in order to most clearly impart an understanding of its operation to the reader.

Referring to Figured 2 it will be noted that the analog signal from pre-conditioning module 13 is first sampled at 16 kHz by sampler 31 thereby producing a sampled signal. The sampled signal is split 22 ways, each of the split signals providing an input to digital filters 35A-35V which filter the signal into quarter octave frequency bands.

Digital filters 35A-35V are bandpass and logarithmically spaced with the base frequency being at 150 Hz. Each filter is of a 6th order Chebychev Type I bandpass type implemented in three second order sections. The quarter octave frequency bands are as shown on the following page :

Filter Band	Lower Frequency Boundary (Hz)	Upper Frequency Boundary (Hz)
A	150.00	178.38
B	178.38	212.13
C	212.13	252.27
D	252.27	300.00
E	300.00	356.76
F	356.76	424.26
G	424.26	504.54
H	504.54	600.00
I	600.00	713.52
J	713.52	848.53
K	848.53	1009.10
L	1009.10	1200.00
M	1200.00	1427.00
N	1427.00	1697.10
O	1697.10	2018.20
P	2018.20	2400.00
Q	2400.00	2854.10
R	2854.10	3394.10
S	3394.10	4036.30
T	4036.30	4800.00
U	4800.00	5708.20
V	5708.20	6788.20

The bandpass filtered signal from each of digital filters, for example 35A, is connected to an amplitude detection module 37A and a period estimation module 39A. The output AMP[A] of amplitude detection module 37A is a digital signal representing an estimation of the amplitude of the sampled signal from filter 35A. The construction of module 37A is straightforward and will not be explained in detail other than to say that it could be based on a series of comparators.

Period estimation module 39A counts sampling periods between positive zero crossings of the signal from filter 35A. The output signal PERIOD[A] is scaled so that it is expressed in units of "timeslices".

One "timeslice" being the time taken to deliver one stimulation pulse by means of an electrode. With reference to Figure 3E an example of a stimulation current pulse waveform comprises a first "phase" 103 being a square wave of predetermined amplitude, an interphase gap 105 and a second phase 107 being a current square wave of the same magnitude and duration as the first phase but flowing in the opposite direction between an intra-cochlear electrode and (in mono-polar mode) an extra-cochlear electrode. Time periods 110 and 112 are present in which the system generating the stimulations has time to perform any necessary operations in order to configure for the next stimulation. The overall time taken to set-up, deliver and recover from application of a stimulation is one timeslice, in the present example a timeslice is of approximately 69 microseconds duration.

In the present implementation the maximum stimulation rate is 8kHz whereas the sampling rate is 16kHz. Accordingly PERIOD[A] is the number of samples occurring between positive-going zero crossings divided by two and rounded. The PERIOD[A] signal is updated at the end of each period.

The AMP[A],...,AMP[V] and PERIOD[A],...,PERIOD[V] signals contain magnitude and period information concerning the ambient sound picked up by microphone 11 for each of the frequency bands monitored by bandpass filters 35. It is possible to simply stimulate via each corresponding electrode e[i] with a current intensity corresponding to AMP[i] at a time PERIOD[i] into the future in order to convey the information generated by amplitude detectors 37 and period estimators 39 to a wearer of the cochlear prosthesis. For example, with reference to Figures 3A, 3B and 3C a stimulation sequence via electrode e[A] is shown corresponding to amplitude and period values generated by amplitude detection module 37A and period estimation module 39A as shown plotted in Figures 3A and 3B. Period[A] is equal to P1 at time t=0 and Amp[A] is equal to a1. Accordingly at a time t=P1 a stimulation current is delivered via electrode e[A] of electrical amplitude I(a1) where I() is a loudness growth function which maps amplitude into the dynamic range of the wearer of the prosthesis.

The period P2 and amplitude a2 values are then obtained and a further stimulation is delivered at time t=P1+P2 of amplitude I(a2). This process is repeated continuously to produce the stimulation sequence of biphasic current pulses shown in the graph of Figure 3C. As previously mentioned, such a process could be simultaneously performed independently on all channels of the implant, (a "channel" as used here refers to a stimulation electrode, its corresponding filter and period and amplitude estimation modules).

In the system thus far described the period estimation module 39A produces a period estimate which is simply the time delay between the last two positive-going zero crossings. While this system works well, any noise on the individual period estimates will degrade the performance of the system. To prevent this, it is desirable to calculate a smoothed period estimate.

The individual period estimates constitute a number sequence which is amenable to any of the methods of smoothing known to the art of digital signal processing. The smoothing may, for example, be implemented as a simple FIR or recursive digital filter - preferably a low-pass filter. Alternatively a rank-order filter, such as a median filter may be used. A rank-order filter has the advantage that it will completely remove any single data errors from the number sequence. A smoothed period estimate is thus produced by applying the sequence of period estimates to a digital filter, and taking the output of that filter. The smoothed period estimate is then utilised by taking the most recent output from the filter and using it in place of the (unsmoothed) period estimate.

With reference to Figure 3D there is shown a stimulation sequence via electrode e[B]. It will be noted that stimulation pulse 102 occurs at exactly the same time as stimulation pulse I(a2) of Figure 3C. Such coincidences occur more and more frequently depending on the number of channels used.

There are at least two problems associated with the above stimulation strategy by which stimulations may be delivered by two or more electrodes simultaneously. Firstly, it is well known that simultaneous, or very near simultaneous, stimulation produces a deterioration in the quality of the sound perceived by the user.

Consequently the application of stimulations on a number of channels simultaneously is undesirable. A further problem is that simultaneous stimulation requires very substantial processing power and so is not possible in the majority of cochlear implants presently available.

In light of the above problems the inventors have incorporated a preferred refinement for determining which information signals, coming from amplitude detectors 37 and period estimation modules 39 are most appropriate for acting upon in order to produce a high quality percept in a user. According to the invention, for any stimulation period i.e. "timeslice" t0 the signals AMP[A,t0],...,AMP[V,t0] are ordered according to magnitude and a stimulation current is generated by means of the electrode which corresponds to the signal AMP[A,t0],...,AMP[V,t0] having the greatest magnitude. (It should be noted that, when implanted, electrodes e[A],...,e[V] are tonotopically mapped to filters fA,...,fV so that electrode e[A] is most apically placed whereas electrode e[v] is most basally placed.) For example, if it is found that AMP[F,t0] has the greatest magnitude at a given timeslice then electrode e[F] is used to deliver the monopolar stimulation in the next timeslice t1.

A further variation to this scheme is that a number of the next largest magnitude signals are also determined in the same timeslot, for example AMP[G,t0] > AMP[B,t0] > AMP[K,t0] might be determined to have the magnitudes next greatest to AMP[F,t0]. In that case those values are assigned to AMP[G,t1], AMP[B,t1] and AMP[K,t1] respectively. During the next timeslice, t1, the procedure is repeated and it may be that AMP[G,t1] is selected as having the greatest magnitude so that electrode e[G] is selected for delivering a stimulation pulse of amplitude corresponding to AMP[G,t1]=AMP[G,t0]. By using this scheme it is possible that signals having a large magnitude, though not the greatest, are presented to the user after a short time delay. The inventors have found that most acoustic power is centred around the lower frequency bands which have longer periods associated with them whereas the higher frequency bands generally have less power associated with them as well as being of shorter period. Accordingly, it is predominantly higher frequency sounds which are delayed rather than lower frequency sounds.

A further refinement is that rather than calculate period estimates in respect of the outputs from filters centred at higher frequencies, for example for filters FI,...,FV period estimators 39I,...,39V simply generate a constant signal, or periodicity value, indicating a period of 1.25ms i.e. a periodicity value towards the highest stimulation rate that the device is capable of supporting.

While the preceding description covers a system utilising period estimators on some or all of the bandpass filtered signals, it is possible to implement the system more simply. A stimulus could be requested each time a positive zero-crossing is detected on a filter output. Once per timeslice each channel is interrogated to see if it has a stimulation request pending. If there are no requests pending, then no action is required. If there is exactly one request pending, then a stimulus is generated corresponding to that request.

If more than one request is pending, then the following actions are taken. The requests are sorted according to the amplitudes of the corresponding bandpass filtered signals. A stimulus is generated corresponding to the bandpass filtered signal with the largest amplitude. The next N largest (preferably 5 largest) amplitude requests are delayed by one timeslice. Any remaining requests are cancelled.

This system is simpler to implement than that previously described. It has two main disadvantages, however. The previously described system utilising period estimates acts to limit the stimulation rate on higher frequency channels. This is directly beneficial in that excessive stimulation with little information content is avoided. More importantly, in the case of relatively large amplitude high frequency signals, the lower frequency signals will be completely lost in the simpler system. The rate-limiting effect of the period-estimation system will mean that the low-frequency signal will always get through.

The request generators 41-A,..,41-V and request arbitrator 43 operate to determine which electrode will be stimulated from the AMP[A],...,AMP[V] and PERIOD[A],...,PERIOD[V] signals. The operation of the request generators and the request arbitrator, in order to implement the aforedescribed scheme, will now be explained with exemplary reference to request generator 41A.

The AMP[A] and PERIOD[A] signals are inputs to request generator module 41A. Another input to the request generator is the CLK signal which corresponds to the present timeslice. The CLK signal is modulus 256 to avoid overflow problems. The last input to request generator 41A is a command signal ARB_CMD[A] from request arbitrator 43.

The outputs from request generator 41A are TSLICE[A] and REQ_AMP[A].

The TSLICE[A] represents the time at which it is proposed by generator 41A that a stimulation be delivered having an amplitude value represented by REQ_AMP[A].

The relationship between TSLICE[A] and PERIOD[A] and REQ_AMP[A] and AMP[A] is determined by the value of the ARB_CMD[A] signal. The ARB_CMD[A] signal can take one of three values REQUEST, DELAY, DISCARD. When ARB_CMD[A] takes the value :

REQUEST

then REQ_AMP[A] := AMP[A]; TSLICE[A] :=CLK+PERIOD[A]

DELAY

then TSLICE[A] := TSLICE[A] + 1

DISCARD

take no action.

The principle behind these rules is that in the event that request arbitrator 43, whose operation will shortly be described, determines that a stimulation pulse should be applied corresponding to the output from filter 35A then by sending an ARB_CMD[A] signal having the value REQUEST to request generator 41A the amplitude and timing of the stimulation pulse is made available at the next timeslice. Alternatively, if ARB_CMD[A] takes the value DELAY then the corresponding TSLICE[A] variable is incremented. Construction of the request generator, in order to implement the above rules is readily accomplished according to established synchronous digital design techniques.

Request arbitrator module 43 takes as its input the signals TSLICE[A],.., TSLICE[V] from each of the request generators 41A-41V, REQ_AMP[A],.., REQ_AMP[V] and the CLK signal. Arbitrator module 43 generates a signal P_CHAN which identifies which of electrodes e[A],..,e[V] of the electrode array is to be used to apply a stimulation.

The arbitrator module also generates a signal P_AMP which takes a value REQ_AMP[A],..,REQ_AMP[V] which is used, after scaling as will subsequently be described, to determine the amplitude of the signal to be used when applying stimulation. Request arbitrator module 43 operates according to the following rules:

1. Find all TSLICE[i] with a value equal to CLK.
2. Find N channels of those determined in Step 1 which have the largest value of REQ_AMP[j]. The channel with the largest value of REQ_AMP[j] and TSLICE[j] as determined in step 1 is found and P_CHAN set to j and P_AMP set to REQ_AMP[j]. So that a stimulation is directed via electrode e[j] with amplitude scaled from the value P_AMP=REQ_AMP[j]. This is accomplished by setting the ARB_CMD[j] signal to REQUEST.
3. The channels having the next N-1 largest amplitude values REQ_AMP[] are delayed by one timeslice for consideration during set up for the next stimulation. This is achieved by sending an ARB_CMD[] signal to the corresponding N-1 request generators to DELAY.
4. The remaining channels, which were selected in step 1 but not in step 2 are discarded. This is achieved by sending the corresponding request generators an ARB_CMD[] signal of value DISCARD.
5. If any of the request generators is sending a specific "no pulse request" then the corresponding ARB_CMD[] signals are set to REQUEST.

Once the P_CHAN and P_AMP values have been determined they are passed to Loudness growth function module 47. The growth function module takes into account the predetermined comfort and threshold levels of the user of the cochlear prosthesis in order to map the P_AMP values into the listeners dynamic range. Such mapping is known in the prior art and the reader is referred to US Patent No. 4,532,930 to the same applicant for further details.

The invention is most conveniently practised, in accordance with the embodiment of Figure 1, by programming a SPrint speech processor, available from Cochlear Limited of 14 Mars Road, Lane Cove, NSW 2066, Australia, in order to perform the operations described in relation to Figure 2. The SPrint speech processor is used in conjunction with a CI24M cochlear implant receiver-stimulator from the same vendor.

With reference to Figure 4 there is depicted a block diagram of the overall operational procedure for implementing the present invention in software. Figure 5 further details the operational steps performed in box 201 of Figure 4.

While the invention has been described with reference to preferred embodiments, it is to be understood that these are merely illustrative of the application of principles of the invention. Accordingly, the embodiments described in particular should be considered exemplary, not limiting, with respect to the following claims.

THE CLAIMS DEFINING THE INVENTION ARE AS FOLLOWS:

1. A cochlear implant prosthesis of the type having a transducer for converting an acoustic signal to an electrical signal, a plurality of bandpass filtering means responsive to said electrical signal and operatively producing a plurality of bandpass filtered signals, signal processing means responsive to said plurality of bandpass filtered signals operatively generating stimulation commands, electrode driving means responsive to said stimulation commands and an electrode array coupled to said electrode driving means for operatively delivering to a user of said cochlear implant prosthesis stimulations in accordance with said stimulation commands, said signal processing means including:

a) period estimation means, responsive to said bandpass filters operatively generating periodicity signals indicative of the periodicity of a number of said plurality of filtered signals;

b) amplitude estimation means responsive to said bandpass filters operatively generating magnitude signals indicative of the magnitude of each of said plurality of filtered signals;

c) selection means responsive to said magnitude signals arranged to determine a filtered signal of said plurality of filtered signals having a greatest magnitude, said selection means generating said stimulation commands including a command to stimulate by means of an electrode operatively tonotopically best corresponding to said filtered signal, said command to stimulate further specifying a time for stimulation to occur dependent on a corresponding one of said periodicity signals.

2. A method of operating a cochlear implant prosthesis of the type including a plurality of bandpass filters each having a characteristic centre frequency, said filters generating a corresponding plurality of filtered signals, said prosthesis further including stimulation delivery means coupled to an electrode array, said method including the steps of :

a) in each of a number of time intervals determining the amplitude and a periodicity value for each of said plurality of filtered signals;

b) determining which of said plurality of signals has the greatest magnitude;

c) applying a stimulation current by means of an electrode of said electrode array tonotopically closest to the centre frequency of the bandpass filter producing the signal determined in step b), said stimulation current being applied during a time interval determined from the periodicity value of the signal determined in step b).

DATED this 13th day of July, 1999

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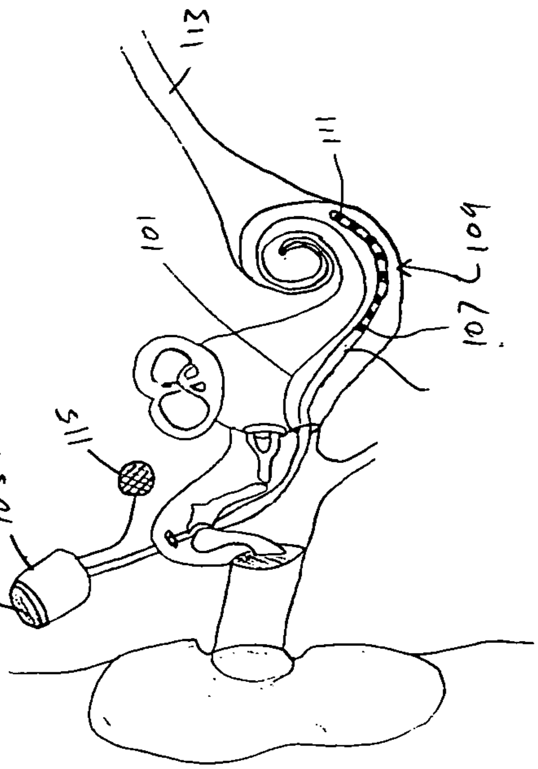

Fig 1

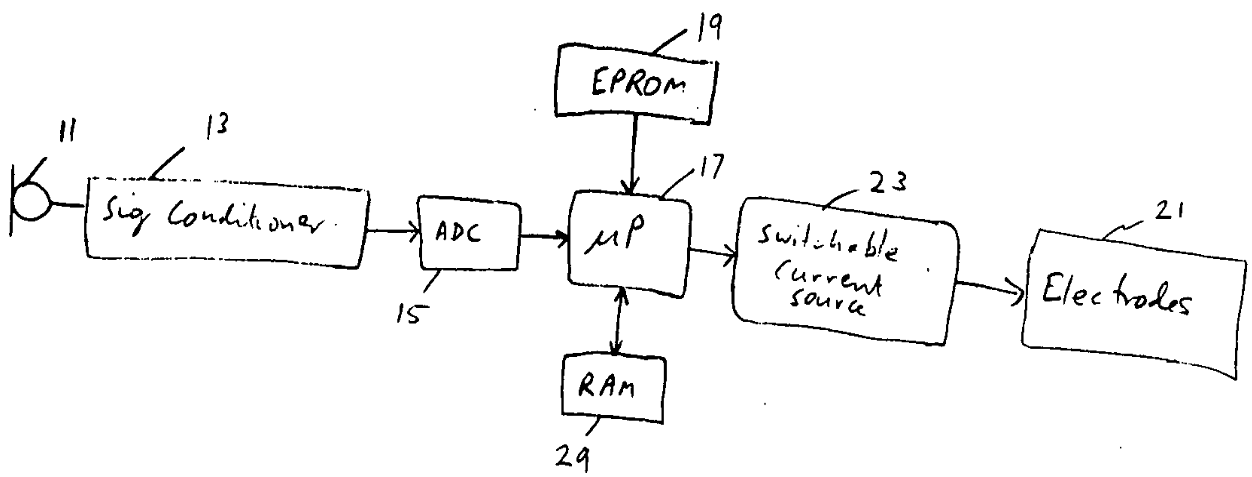
11
13
Sig Conditioner
ADC
15
19
EPROM
17
μP
RAM
29
23
Switchable
Current
source
21
Electrodes

Fig 1A

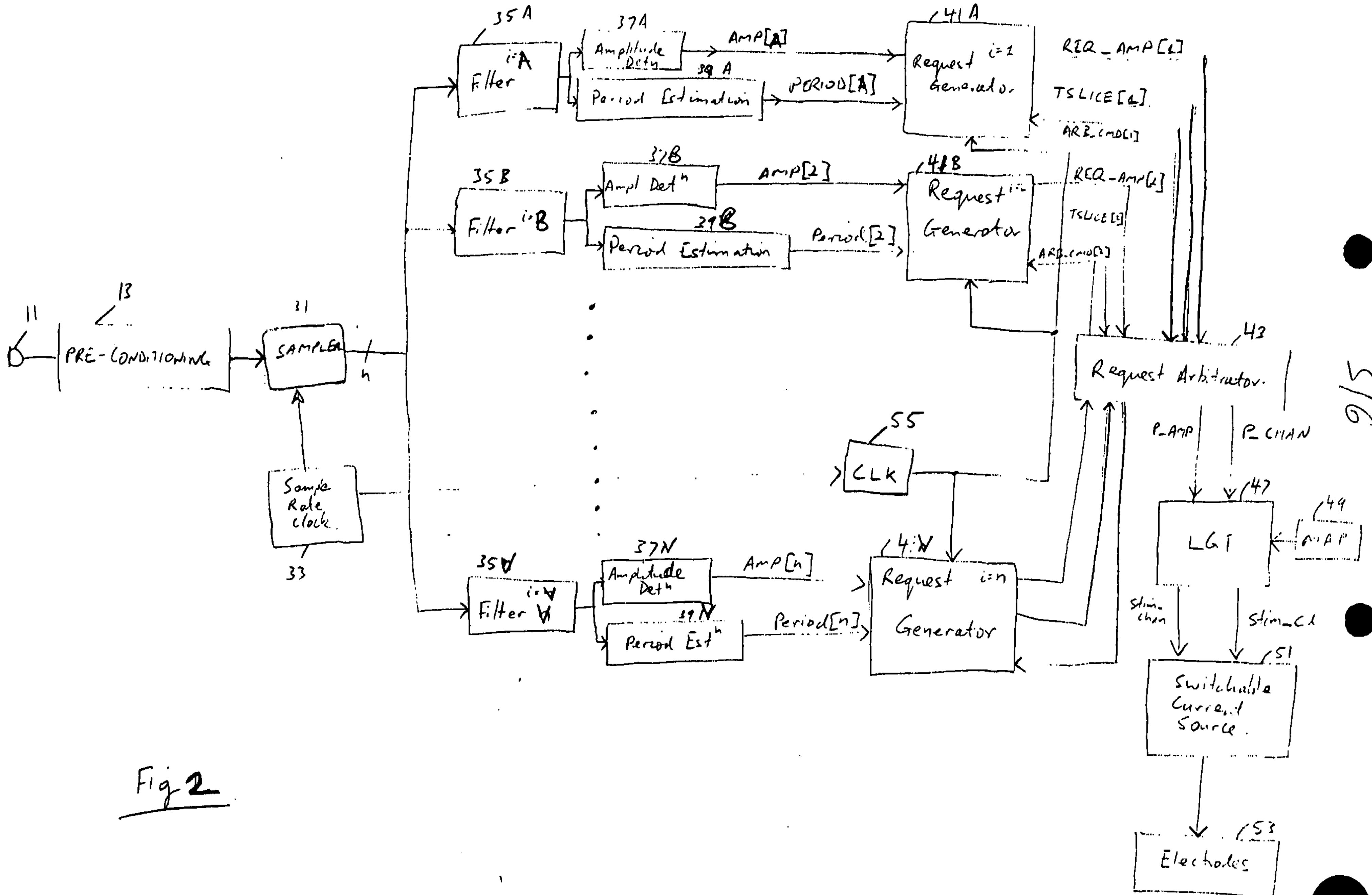
11
13
PRE-CONDITIONING
31
SAMPLER
n
Sample Rate clock
33
35A
Filter i:A
35B
Filter i:B
35N
Filter i:N
37A
Amplitude Det^n
39A
Period Estimation
37B
Ampl Det^n
39B
Period Estimation
37N
Amplitude Det^n
39N
Period Est^n
AMP[A]
PERIOD[A]
AMP[2]
Period[2]
AMP[n]
Period[n]
41A
Request i=1 Generator
41B
Request i:2 Generator
41N
Request i=n Generator
55
CLK
REQ_AMP[1]
TSLICE[1]
ARB_CMD[1]
REQ_AMP[2]
TSLICE[2]
ARB_CMD[2]
43
Request Arbitrator
P_AMP
P_CHAN
47
LGI
49
MAP
Stim_chan
Stim_Cl
51
Switchable Current Source
53
Electrodes

Fig 2

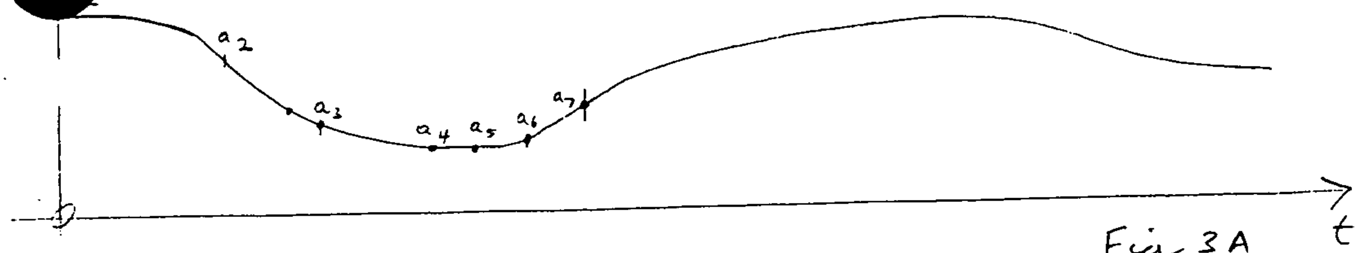

Fig 3A

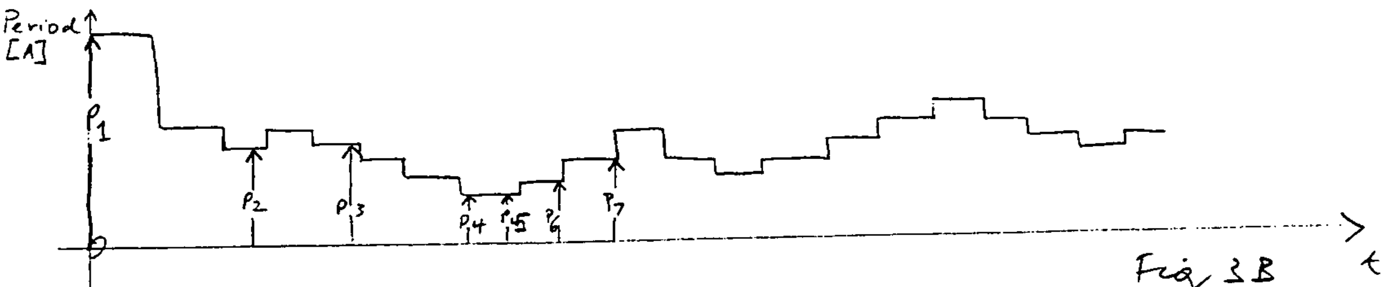

Fig 3B

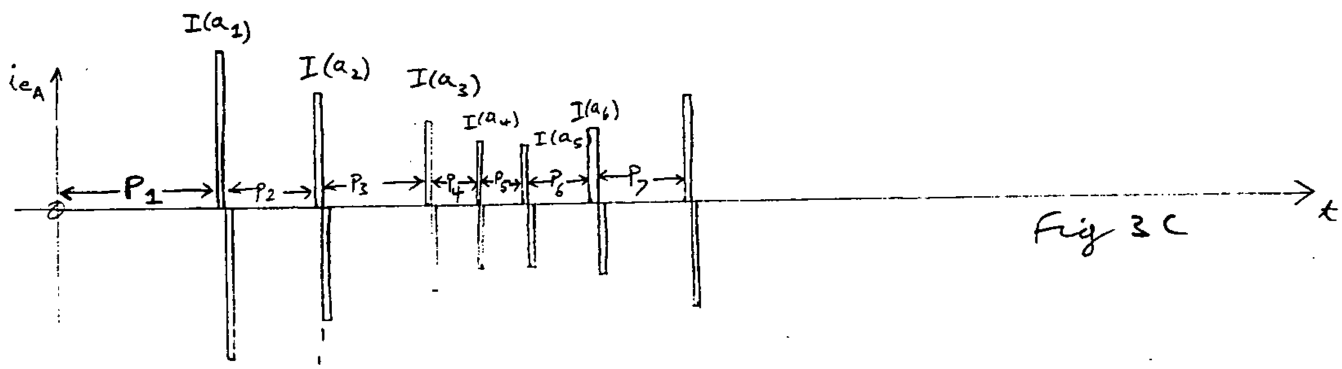

Fig 3C

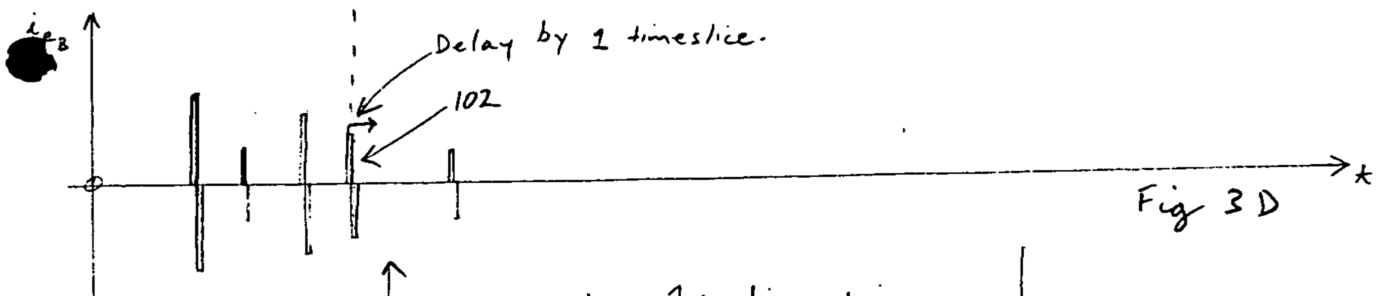

Fig 3D

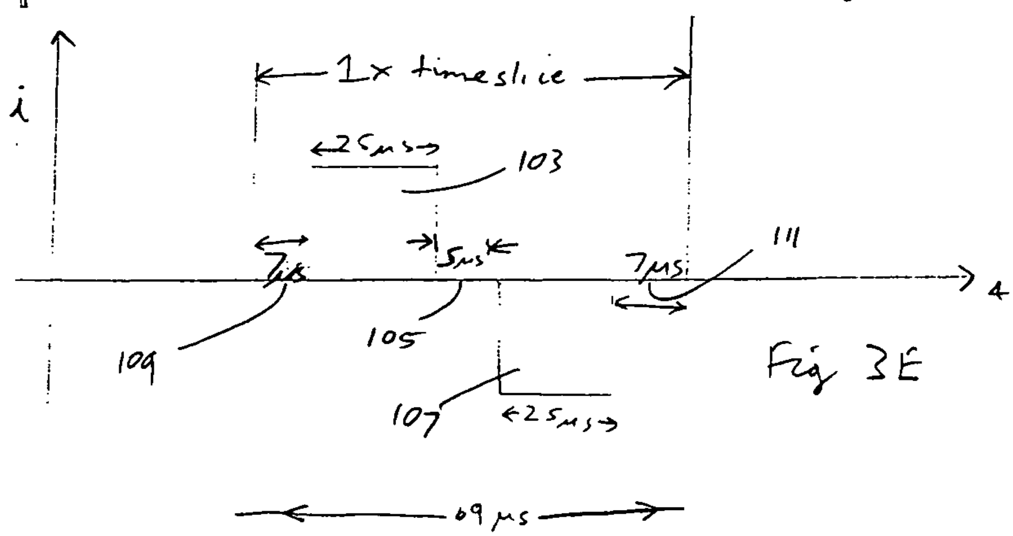

Fig 3E

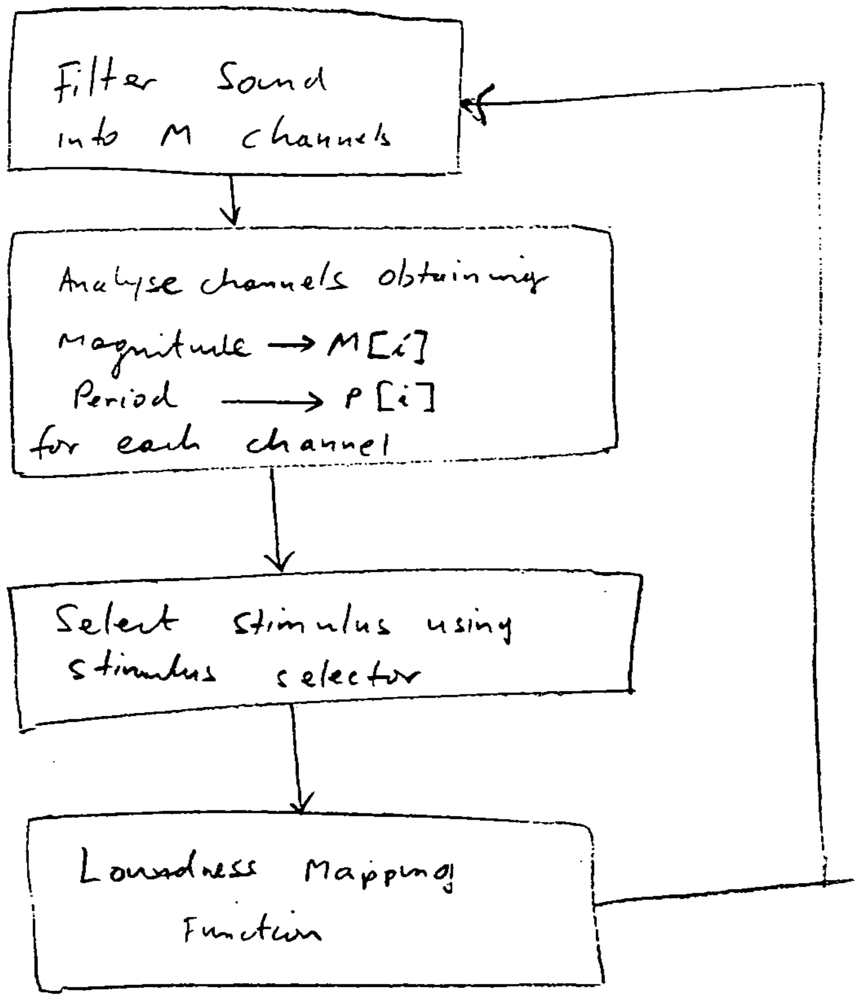
Filter Sound
into M channels
Analyse channels obtaining
Magnitude → M[i]
Period → P[i]
for each channel
Select stimulus using
stimulus selector
Loudness Mapping
Function

Fig 4

Initialisation

Update all channel inputs to stimulus selector

$A[i] = M[i]$, $T[i] = CLOCK + P[i]$

Start stimulus selector

find all channel inputs requesting immediate stimulation ie

$T[i] = clock$

Sort channel inputs using $A[i]$

Delay the 2nd largest channel through to the (N-1)th largest channel

Delay channel j by:

$T[j] = T[j]$ + minimum stim time

Stimulate the largest magnitude channel.

Update channel j by:

$A[j] = M[i]$, $T[j] = CLOCK + P[i]$

Discard the remaining channels by updating the channel inputs.

Update channel j by

$A[j] = M[i]$, $T[j] = CLOCK + P[i]$

Fig 5